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***Abstract*—We propose a method to accelerate the acquisition of High Angular Resolution Diffusion Magnetic Resonance Imaging (HARDI) in order to promote its application in a clinical setting. The method relies on a Spherical Deconvolution approach where the fibre orientation distribution (FOD) is recovered in all voxels simultaneously. The dMRI acquisition is meant to be accelerated through the partial Fourier sampling of each diffusion weighted image. Despite the further reduction of the acquired information, the FOD estimation still preserves its angular resolution thanks to the structured sparsity prior that is imposed in the problem.**

Diffusion magnetic resonance imaging is a unique non-invasive technique, enabling to extract information about the microscopic structure of white matter tissue in vivo. Spherical deconvolution approach is one of the several methods that have been developed in order to extract the fibre orientation information at high angular resolution. However, they rely on, at least, 60 diffusion images leading to considerable long acquisition times that prevent their application in the clinical setting.

The present study is based on the works of Daducci [1] and Auria [2], which exploit the recent theory of Compressive Sampling in order to recover the fibre orientations from a reduced number of diffusion images (q-space sampling). In particular, the work of Daducci. promotes the fibre orientation distribution sparsity through a voxel-wise ℓ_0-minimisation, suggesting an accurate reconstruction from no more than 30 q-space images. Auria et al. have built on this approach and introduced a spatial regularization prior promoting the smoothness of the spatial variation of fibre orientations, suggesting the FOD reconstruction from no more than 15 q-space images. We propose an extension of the above-cited methods where the acquisition of each diffusion image is accelerated through partial Fourier (k-space) sampling in order to fully exploit the spatial regularisation prior of Auria et al..

A novel linear measurement model is defined, mapping the matrix $X \in \mathbb{R}^{n\times N}$, which represents the FOD in each voxel of the imaged brain, onto the kq-space samples $\hat{Y} \in \mathbb{C}^{(N_c\times N_g)\times k}$ as follows:

$$\hat{Y}_{q,c} = M_q F P X M S_0 C^{(c)} H^{(q,c)} F M_k^{(q)} + \eta_{q,c} \qquad (1)$$

Each line of $\hat{Y}$ corresponds to the sub-sampled k-space of the DW-image acquired with gradient q by the channel with sensitivity c. F represent the Fourier matrix, the matrices $M_q \in \mathbb{R}^{1\times n}$ and $M_k \in \mathbb{R}^{N\times k}$ are binary masks representing the joint kq-space under-sampling of interest. The columns of the matrix $P \in \mathbb{R}_+^{n\times n}$ represent the ensemble average propagator for a single fibre oriented in all possible n directions. Voxels outside the brain are modelled with zero-signal through a diagonal matrix $M \in \mathbb{R}^{N\times N}$ while the diagonal matrix $S_0 \in \mathbb{R}^{N\times N}$ stores the intensities of the non-diffusion weighted image. The acquisition of the diffusion signal from multiple channels is taken into account through the diagonal matrix $C^{(c)} \in \mathbb{C}^{N\times N}$ which stores the sensitivity map of channel c. Motion and magnetic field inhomogeneities generate a phase distortion that is accounted in the matrix $H^{(q,c)} \in \mathbb{C}^{N\times N}$. The multi-channel sensitivities are assumed to be estimated from the non-diffusion weighted image while the phase distortion can be calibrated from low-resolution diffusion weighted images. Measurements $\hat{Y}_{q,c} \in \mathbb{C}^{1\times k}$ are assumed to be contaminated by Gaussian noise $\eta_{q,c} \in \mathbb{C}^{1\times k}$.

The fiber orientation distribution in each voxel is recovered solving a minimisation problem of the following form:

$$\min_{X\in\mathbb{R}_+^{n\times N}} \|\mathcal{A}(X) - \hat{Y}\|_2^2 \quad \text{subject to} \quad \|K_d \cdot X \cdot K_v\|_0 \leqslant \kappa \qquad (2)$$

where $K_d \in \mathbb{R}^{n\times n}$ and $K_v \in \mathbb{R}^{N\times N}$ represent two blurring bases imposing correlation within neighbour directions and neighbour voxels respectively, while $\mathcal{A}(\cdot)$ is the linear operator acting on X and modelling the measurements $\hat{Y}$. The parameter κ acts as a bound on the sparsity of X and it is computed as the number of voxels times the average number of fibre expected per voxel. We propose a multiple-shell approach for the q-space sampling in order to gain more accurate identification of the white matter tissue, joint with a uniform random k-space sampling with a fully sampled low frequency zone.

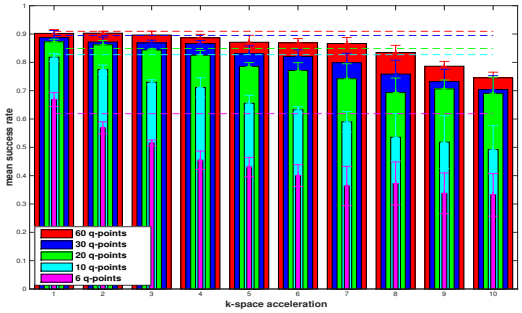

Fig. 1. Mean success rate index evaluating the performances of the FOD reconstruction in the case of different q-space and k-space undersampling settings. The results have been obtained from synthetic data with SNR=30.

The fibre orientation reconstruction has been tested through numerical simulations and real data in presence of different acceleration rates. The results suggest that the kq-space approach can significantly outperform the q-space sampling up to an acceleration of 7 with 60 diffusion images in simulated data (see Figure 1), rising the hopes to open the door of clinical applications for high angular resolution diffusion imaging methods.

References

[1] A. Daducci, D. Van De Ville, J. P. Thiran, and Y. Wiaux, "Sparse regularization for fiber odf reconstruction: From the suboptimality of ℓ_2 and ℓ_1 priors to ℓ_0," *Medical Image Analysis*, vol. 18, pp. 820–833, 2014.

[2] Auría, A. and Daducci, A. and Thiran, J.-P. and Wiaux, Y., "Structured sparsity for spatially coherent fibre orientation estimation in diffusion mri," *NeuroImage*, vol. 115, 2015.